Insulating Biomaterials
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Introduction to Insulating Biomaterials for Implantable Electronics

Microelectronic components and integrated circuits have been used extensively in cardiac pacing and cochlear prosthesis systems. Both of these systems utilize large titanium or ceramic canisters for packaging the electronics, and glass feedthroughs for interconnecting with stimulation leads and electrodes.

The pace of development of these implantable devices has paralleled the pace of technological innovation. Development of the cardiac pacemaker is one good example. The first recorded demonstration of electrical pacing of the heart occurred in 1872 using a system consisting of a 300V battery and hand held electrodes [1]. Following that important demonstration, it wasn't until 80 years later when endomysial electrodes, transistors, epoxy encapsulation, and high energy density batteries became available that the stage was set for demonstration of the fully implantable cardiac pacemaker in the 1950s [2]. Lithium battery technology, improved packaging, and improved lead technology have recently resulted in complex pacemaker systems with implantable lifetimes of up to 15 years [3].

Today's pacemakers consist of hermetically sealed titanium canisters with glass sealed feedthroughs. The size of the pacemaker is dominated by the size of the battery and the titanium case which is a few centimeters in diameter and a half centimeter thick. Modern leads are typically multi-filament coils of a high nickel content stainless steel alloy (MP35N).

Throughout the history of the pacing industry, implantable materials considerations have dominated technological innovation. Silicone insulated leads have been very reliable and have been used for decades in cardiac pacing applications. Other lead insulation materials have been used to overcome the tendency of silicones to stick to tissue during insertion, and to reduce the diameter of the pacing lead [3]. Polyurethanes used for this application were marketed in the 1970s. However, in several long term clinical studies, polyurethane leads from several manufacturers tragically exhibited substantially higher failure rates than silicone leads [4, 5]. A possibly overlooked clue that may have alerted pacing lead designers to possible long term instability issues with polyurethanes was the 1979 observation that both polyether and especially polyester urethanes were susceptible to hydrolytic degradation during relatively short term water soak [6].

The most important goal of current work on insulating biomaterials is to identify possible failure modes for insulating biomaterials to avoid such clinical surprises.

Many applications for implantable microelectronics and micromachined devices are under development including biochemical sensors and drug delivery systems. Most of the applications require only subcutaneous implantation so the traditional titanium canister is the package of choice.

Neuroprostheses

Micromachined neuroprostheses, however, have more demanding applications. Neuroprostheses are being developed for rehabilitation of the deaf, blind, spinal cord injured, and amputees. Most of the neuroprosthesis concepts require close proximity to

the small (10μm nominally) and fragile cells of the nervous system. Many will be attached or embedded directly in neural tissue. Not only is neural tissue tightly packed with functional elements, but it is dynamic. Peripheral nerves stretch and relax with every motion of a limb. The spinal cord moves within the spinal canal stretching the spinal roots with every bend. The brain moves relative to the skull and itself with every heartbeat, breath, and motion of the head. Eyes are constantly in motion creating substantial forces of acceleration on the retina.

Requirements Peculiar to Many Neuroprostheses

Because of the biological environment, the fragile nature of neural tissue, the tight packing density of neurons, the relative motions of tissues, and the effects of acceleration, an implantable electronic device for neuroprostheses must have the following characteristics to be clinically successful:

1. BIOCOMPATIBILITY
2. BIORESISTANCE
3. SMALL SIZE
4. DENSITY MATCHED TO NEURAL TISSUE
5. MINIMALLY TETHERED TO ADJACENT STRUCTURES

Biocompatibility is essential to minimize formation of connective tissue between neurons and electrodes over the course of long term implantation. Bioresistance is essential to ensure that the implanted electronic device remains functional for decades. Small size (measured in micrometers) is necessary to minimize damage to target neural structures during placement and to allow devices to fit into small spaces without damaging adjacent tissue. Matching the density of the device to neural tissue is important to avoid damage to the device or tissue from differential acceleration caused by intrinsic motion of the tissue, as well as from extrinsic accelerations caused by everyday living or traumatic incidents. Minimal tethering reduces the likelihood that forces will be transmitted from wiring to the implant.

The ideal implantable neuroprosthetic device would thus have: no chemical reactivity (an important aspect of both biocompatibility and bioresistance); zero dimensions to avoid traumatic damage to the target and surrounding tissues; no mass in keeping with no size to avoid acceleration damage; and no tethering. Devices approximating this ideal are being developed at a variety of locations worldwide using micromachining of silicon as a primary tool. Silicon is a good choice since it readily forms an inert, self limiting oxide that is biocompatible [7-9]. Micromachining can be used to produce a variety of novel structures that may approximate the ideal neuroprosthesis [7-18].

Specific Packaging Issues for Neuroprostheses

As fabricated, micro-machined devices typically are not bioresistant. **Figure 1** illustrates some possible failure modes when an integrated circuit micro-device is immersed in saline environments for extended periods of time.

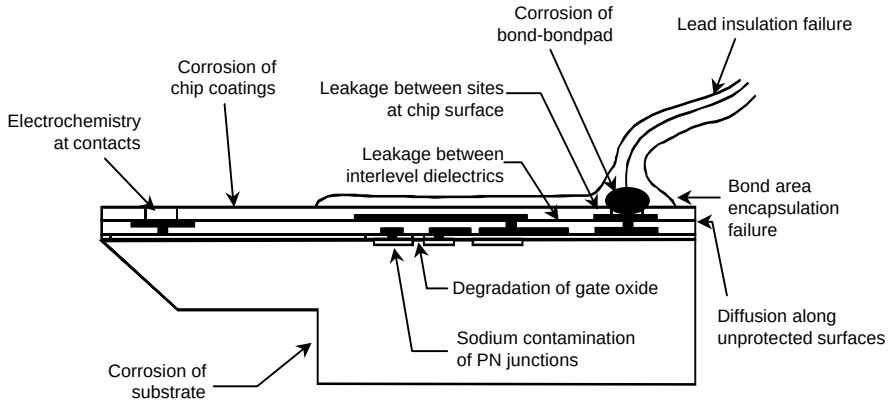

Figure 1: Sketch illustrating potential problem areas for implantable integrated circuits protected by thin layers of insulating biomaterials.

There are several problem areas associated with operation of micro-fabricated electrical devices within physiological systems:

1. The wires used to attach to the device must be capable of withstanding immersion in ionic fluids with a 5 volt magnitude bias across the insulation.
2. The exposed areas where the wires are attached to connectors or devices must be coated with a material that is typically applied after bonding has been accomplished.
3. If microribbon cable technology is used, it is necessary create a void free seal in the area under the microribbon where it attaches to the device.
4. The wire assembly must be flexible enough to minimize static forces applied by the implanted element on the tissue. Dynamic flexing is unlikely to be a problem in the protected areas where the devices are to be used.
5. The circuits on the chip must be protected from water and ionic contamination (i.e. the surface density of sodium ions in the MOS gates should be less than $10^9/cm^2$.
6. The chip surface materials, the chip substrate, and encapsulants must all be bioresistant and biocompatible.

These micro-machined devices will not find useful clinical applications until suitable insulating biomaterials and packaging techniques have been identified for protecting the devices from the environment of the body for many decades.

Common Pitfalls in Evaluation of Insulating Biomaterials

While there are many, many coatings that could be used for insulating electronics and protecting micromachined integrated circuits, few are suitable for direct immersion into biological systems or even aqueous environments. Common materials used successfully by the electronics industry for protecting integrated circuits are not stable in the salt water environment. This fact is often under-appreciated. Several examples exist of unexpected failures from materials that were not tested under realistic conditions. One example was the extensive use and proposed use of polyimide for development of neuroprostheses. This choice was based on the experience of the

electronics industry, but the material had never been fully evaluated under saline conditions. However, since 1971 it has been known that polyimide is susceptible to hydrolysis [19]. Polyimide has also been implicated in feedthrough glass corrosion due to electrolyte pooling under a polyimide coating used in the early 1980s. This defect resulted in the recall of 4,500 pacemakers. Parylene has also been widely promoted as an insulating biomaterial. While effective for short term encapsulation, upon exposure to high humidity, Parylene undergoes hydrolysis with resultant craze cracking [20].

There are many other examples of packaging surprises, some with tragic loss of human life, and all very costly to the industry. The important point is that life testing of materials is an art rather than a science. The tragic mistakes are made when too much emphasis is placed on theoretical models of accelerated testing without due consideration for the assumptions.

The approaches outlined in this proposed work attempt to make judicious use of accelerated testing techniques while also testing at least some of the assumptions made for those tests. More importantly, accelerated testing is being used to point out possible failure mechanisms, but long term, high sensitivity tests to be conducted in parallel with the accelerated tests to ensure that other factors are not being missed. For example, it is always possible that the acceleration factor may also accelerate a repair or stabilization phenomenon which could cover up a degradation process under the accelerated test condition, but not under normal conditions. Likewise, under non-accelerated conditions, there may be parallel paths of leakage currents such that a stable, relatively high but acceptable leakage current masks a lower level current that is steadily increasing with time. Until the lower level leakage pathway becomes comparable to the higher leakage path, it will not be noticed. Merely fitting trends to this data will show early in the process that the device may last indefinitely “As Far As Is Known” (AFAIK). However, if observed long enough, or if the process can be thermally accelerated without changing the nature of the process, then eventually the formerly rock steady leakage current observations may suddenly, and unexpectedly change.

Current Neural Probe Technology

Several groups currently produce micro-machined neural probes for possible use in neuroprosthetics. Stanford uses novel plasma etching techniques to produce highly refined two dimensional structures (G. Kovacs, Stanford and Dave Kewley, Cal Tech personal communications). The University of Utah produces two dimensional arrays of penetrating electrodes using silicon micro-machining [11, 21-23]. At the University of Michigan Center for Neural Communication Technology, many designs of neural probes are available etched from silicon using differential chemical etching techniques [11, 21-23]. Current neural probe technology has a shank sizes on the order of tens of microns. Thin film protective coatings on the order of 1 micron thick for the probe shank would be advantageous to maintain the small sizes. Typically, silicon dioxide and silicon nitride are used as dielectrics for these devices. At MIT, a silicon dioxide/silicon nitride composite dielectric was used for insulation of the polysilicon leads which connect the recording contacts to bonding pads on the periphery [8, 9, 24,

25]. The silicon dioxide layer can be up to 1 micron thick and provides an excellent dielectric to slow the anodization of the silicon nitride overcoating [26, 27]. The silicon nitride layer is usually about 1,500 angstrom thick. Its purpose is to prevent water and sodium migration into the silicon dioxide layer which would allow strong anodization currents to flow in the silicon nitride layer. Together, these coatings provide excellent short term protection for the passive probe technology currently in use.

Active probes (micromachined silicon electrode arrays with integrated circuit stimulators and recording systems) are also under development at the University of Michigan. These devices will allow reducing the lead count to a minimum, while increasing the number of channels of neural information that can be recorded or activated. Prototype devices have been fabricated and used to record and stimulate in short term tests [28]. These devices are based on CMOS integrated circuit technology. Once fully functional, these devices will need to be protected with insulating biomaterials. CMOS technology is particularly suitable for implantable neuroprostheses and other medical devices because very low power circuits can be developed. In addition, CMOS transistors inherently have extremely low input bias current requirements which make them ideal transducers for acquiring bioelectric signals from high impedance microelectrodes.

Unfortunately, low power CMOS circuits require the presence of high impedance circuit nodes and close transistor threshold matching. Thus, subtle changes in the electrical isolation between layers and between conductors on the same layer may cause substantial changes in the operation of the circuits. In addition, small changes in the threshold voltages for the transistors can cause offsets that when amplified exceed the dynamic range of the system. These factors are challenging for packaging even when CMOS devices are not immersed in saline solutions. In saline solutions, the many materials present in a CMOS integrated circuit may interact electrochemically to produce new failure modes not seen in testing of passive technology. In addition, diffusion of species will likely be an important consideration. Diffusion of water molecules could hydrate dielectrics and cause reduced resistivity. Diffusion of ions such as sodium could cause drift of thresholds. Such issues need to be addressed before implantable micromachined CMOS integrated circuits can become clinically useful tools.

Overall Goal and Approach

The goal of this Insulating Biomaterials contract is to develop and evaluate insulating biomaterials for new packaging techniques that minimize volume and maximize long term reliability for neuroprostheses and other implantable devices. A key word in this statement is "evaluate". As indicated above, the evaluation aspects of this work are daunting. There is no set formula for accomplishing the evaluation in a manner that will necessarily uncover all possible failure modes for long term implants. Since the implants must function reliably for decades, simply monitoring their functionality is insufficient because the time for application of the technology is likely to be much sooner. Thus, a multifaceted approach has been taken which attempts to provide early warning of device changes that could lead to failures, attempts to isolate suspected mechanisms of failure, and attempts to accelerate suspected or unknown mechanisms.

Careful interpretation and presentation of the data is crucial to providing the community with reliable information to avoid misapplication or improper extrapolation of the information developed here.

Summary of Current Status of Research on Insulating Biomaterials

This summary of current work is presented to provide perspective for the objectives of the current contract.

Instrumentation

High sensitivity computerized instrumentation and protocols for monitoring leakage currents and resistances of insulating biomaterials has been developed for *in-vitro* work. A wide dynamic range single channel system and a 384 channel narrow range system were developed and are in use. The 384 channel system was constructed as 3 modules to allow setting temperatures for devices under test to 37°C, 80°C, and 90°C for temperature acceleration studies. Methods for assembling devices into long reflux tubes permits long term testing without problems associated with aging and degradation of the caps and connectors.

In-vivo testing was made possible by construction of a sensitive 4 channel portable electrometer system. Devices and materials that showed promise during *in-vitro* testing were implanted subcutaneously and/or subdurally in rabbits. A miniature current limited battery pack was developed to maintain 5 volt bias on the devices between measurements. Measurements were relatively infrequent due to the fragile nature of the percutaneous connections, but even so, many were lost during measurements over the years.

To avoid the long term percutaneous connector problem, an ultra-low power, multi-channel electrometer measurement system was developed for implantation to transmit femtoamp to microamp leakage current information from biased test devices. The first design of this system was focused on evaluation of critical dielectrics, junctions, and transistor thresholds for the CMOS technology being used. Only one external device can be attached. All test devices are continuously monitored for continuity of bias and sense lines. The first working devices were recently obtained and are now under test. The system draws less than 4μA of current and operates at 5 volts. Lithium batteries were identified, encapsulated, and have been under soak tests for over 2 years without evidence of corrosion and no reduction in open circuit voltage so far. A simple optical telemeter has been functional in a rabbit for over two years thus far demonstrating that the optical telemetry link is a viable approach to low power data transmission.

Test Structures

A variety of test devices have been developed to investigate different materials and different aspects of materials. Insulated wire loops are a convenient way to initially screen materials that may be commercially available as a wire coating or that we can coat.

Bare silicon wafers are used to screen materials for bulk resistivity. These devices are fabricated by first coating one surface with the material to be tested, and then attaching a wire to the back surface and potting the assembly leaving a 0.25 inch diameter opening over the material under test. These devices are then immersed in saline for long term testing.

Elasticity is monitored using 1 or 2 mm diameter rods of materials of interest in a cyclic pull test. These pull test devices are sensitive to small structural changes in the material that can be caused by cross linking, depolymerization or other factors. The diameter of the devices is small to promote any diffusion limited processes.

InterDigitated Electrodes (IDEs) are used to evaluate protection of the surface interface between an encapsulant and substrate. All of the materials we are testing have at least a native silicon dioxide layer on the surface prior to encapsulation, so we expect that we are always testing the silicon dioxide to encapsulant chemical bond integrity.

Some of the early IDE structures included an 8 point polysilicon resistor structure that allowed long term monitoring of polycrystalline silicon resistors and contacts. Also, several oxide monitors were fabricated in MOSIS which monitored the stored charge on a UV programmable capacitor. This device was to monitor gate oxide leakage currents from a one time programming of a small storage capacitor, but proved difficult to use and interpret in practice.

To allow more thorough investigation of CMOS circuit elements, the CMOS circuit test chip described above included 14 test devices which allows continuous monitoring of the leakage currents through and between all of the CMOS dielectric layers. Test devices included stacked perforated plates for vertical leakage pathways, IDEs for lateral leakage pathways, large area reverse biased diodes, surface plates for bulk leakage pathway, surface IDEs for encapsulant/chip interfacial leakage pathways, MOS threshold testers, and an uncommitted pad set for attachment of an external IDE or other test structure.

Commercial Materials

A wide variety of materials have been evaluated for lead insulation *in-vitro* with only silicones, fluorocarbons, and one polyesterimide exhibiting sufficient longevity for further investigation *in-vitro* and *in-vivo*. Temperature acceleration protocols have been developed in an effort to speed identification of possible sources of long term failure in these materials. Animal testing for up to 2 years and temperature acceleration have further verified the suitability of these materials for long term implantation. Bending tests and bonding tests have shown that fine platinum wire may be the ideal interconnect material since it is more flexible that gold wire of the same tensile strength. However, insulated wire of very fine diameters is not currently available commercially.

Many materials have been evaluated for encapsulation of silicon surfaces for protection of lead bond areas and exposed traces. Only some silicones have exhibited any capacity whatsoever to accomplish this task. Fortunately, some silicones have successfully insulated test structures for over 8 years of continuous biased saline soak.

Temperature acceleration and *in-vivo* testing for over 2 years have further verified the functionality of these silicones. Pull tests of 1mm diameter silicone samples has revealed that while most silicones maintain stable elasticity during long term saline immersion and long term subcutaneous implantation, some rapidly degrade. It appears that the difference may be related to the filler used. The samples that degraded were fabricated with optically clear silicones that utilize resin filler for optical clarity. The samples that maintained their elasticity were fabricated with fumed silica filler. We have also found that the elasticity property is sensitive to cure conditions which may indicate that chemical-structural changes can occur under some conditions. Further investigation of these observations is needed to avoid the pitfalls found during polyimide and polyurethane testing.

Teflon® (mostly PFA but others as well) has exhibited the highest insulation resistance for the longest time of all materials tested. Other than failures related to defects in the films, Teflon® insulated wire exhibits no signs of degradation over many years of *in-vitro* saline soaks at all temperatures. Similar success has been found during animal tests for over 2 years so far. Unfortunately, Teflon® does not adhere to silicon dioxide surfaces, and thus cannot be used to protect the bond area of an implant. However, Teflon® wire can be overcoated with silicone elastomers to form a permanent "o-ring" like seal that seems to perform well indefinitely. Thus a 2 part system with Teflon® insulated wiring and silicone encapsulated bond areas is recommended at the present time for long term animal implant assembly. We have fabricated dozens of structures using this approach and have verified the performance *in-vitro* at the three temperatures and *in-vivo* for over 2 years. However, there may be legal prohibitions for clinical applications due to prior litigations involving use of Teflon® as a structural implant element.

Ceramic coatings such as silicon dioxide and silicon nitride are of particular interest and were extensively tested. Thin films of silicon dioxide become highly conductive when immersed in saline solutions. Silicon nitride, in spite of being a super barrier to ionic diffusions in semiconductor work, dissolves slowly when immersed under bias in saline solutions. However, it was also discovered that inclusion of a thermal oxide layer under the silicon nitride layer greatly slowed the dissolution. Application of a very thin film of silicone or silicon dioxide over the surface of the silicon nitride layer further reduces the dissolution process. This was an important discovery because we were counting on silicon nitride for providing a sodium barrier for the CMOS micromachined probe technology.

New Materials

New materials have been developed using Chemical Vapor Deposition techniques to allow application of thin, conformal, high purity films to micromachined silicon devices. Liquid silicone elastomers and extruded PFA Teflon® demonstrated the appropriate classes of materials that should be used for this application, but they cannot be applied to wires of the dimension appropriate for implantable micro-machined devices. Also, when liquid silicones are applied to bond areas, the resulting structure is more massive than what could be produced by a thin, conformal coating. We have developed several

techniques for deposition of silicones and fluoropolymer by CVD. The films have essentially the same chemistry as the commercially available bulk materials previously tested. They have been conformally deposited on structures as small as the UM micro-ribbon interconnects. While some soak testing has verified that the performance of the CVD films is similar to that of the commercial materials, the reactors used for developing the films were not particularly clean and resulted in inclusions of particulates which compromised most of the testing. However, one set of CVD fluoropolymer coated silicon bulk resistivity testers has exhibited commercial quality resistivity for over 2 years thus far. CVD silicone coated silicon bulk resistivity testers have exhibited essentially bulk resistivities for over 8 months thus far, at 80°C.

Objectives of New Contract

The goal of the work is to identify and evaluate materials, coatings, and assembly techniques suitable for protection of integrated circuit devices being considered for neural prosthetic applications. A lifetime design goal of 100 years of *in-vivo* functionality will provide a reasonable margin of safety for materials defined by this program. A multi-faceted research program is proposed to allow investigation of known failure mechanisms of the materials and techniques under study as well as the discovery of new failure mechanisms. Both *in-vivo* and *in-vitro* testing will be used with a variety of testing procedures, devices, and materials to further discover, develop and understand insulating biomaterials for micromachined devices.

The objectives of *in-vitro* testing are to minimize the use of animals by cost effective rapid screening of materials under consideration, and to identify possible failure mechanisms using accelerated testing by three techniques: 1) sensitive measurement; 2) elevated temperature; and 3) chemical concentration enhancements. *In-vitro* evaluations will be accomplished under conditions similar in simple ways to physiological conditions so that results can be more readily understood in terms of the properties of the materials.

There are two specific objectives for *in-vivo* testing. One is to directly evaluate candidate materials from the *in-vitro* testing in a more realistic test environment for the application - the mammalian brain. A second objective of *in-vivo* testing is to discover new possible failure mechanisms which can then be used to augment the *in-vitro* testing by allowing acceleration of specific biologically related failure mechanisms.

The final objective of the proposed work is to develop and evaluate new materials and application techniques specifically designed for implantable micromachined CMOS integrated circuits useful in neuroprostheses and other medical applications. This portion of the work frees us from always depending on adaptation of materials that are being developed for other applications.

Progress During First Quarter

Instrumentation

Much of our efforts during the first quarter were directed at moving the test instrumentation from the MIT location to InnerSea Technology. InnerSea is a small research company located in Bedford, Massachusetts. There are 1200sq ft of lab space and 1000sq ft of office space. The lab is divided into a 300sq ft clean room, 400sq ft instrumentation room, and 300sq ft electronics development space. Animal work will be accomplished at the West Roxbury VA Medical Facility in a 900sq ft testing laboratory. Animal surgical suites are available within the animal care facility at the VA.

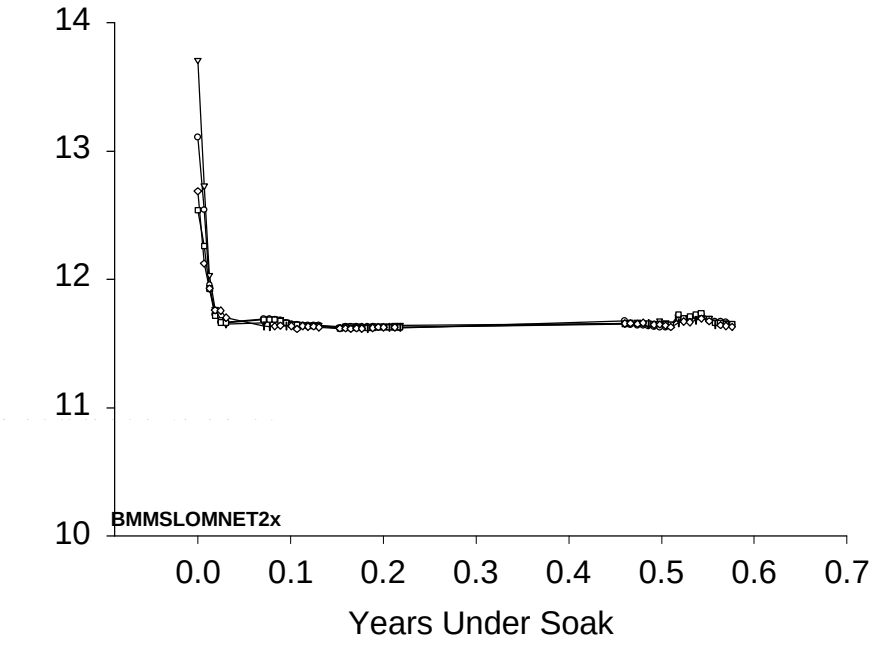

Figure 2: Example data set from Kapton ribbon interconnect assembled with Omnetics connector and Epoxy Technology 353NDT encapsulation of bond area.

The 384 channel electrometer system was successfully moved, repaired, and re-calibrated. During pre-measurement checks, it was determined that several electrometers were malfunctioning. These were repaired, and the measurement system was transferred to a higher speed computer to minimize delays in data acquisition. Each test tube or jar was checked for continuity and saline level. Several assemblies had to have platinum electrodes substituted for the silver reference electrodes previously used because of dissolution of the silver. Devices with relatively high leakage currents had to be moved to lower sensitivity electrometers. All 384 electrometer channels were checked using calibration resistors. An example of readings following the move from MIT are shown in **Figure 2**. The long drop out of data between 0.25 and 0.45 years was caused by the lab fire at MIT. Data taken at InnerSea began after about 0.53 years.

Laboratory computers were networked together to facilitate system monitoring and backup of important data. New temperature controller components were ordered to improve regulation of soak temperatures in several ovens.

Mechanical Properties of Silicones

The mechanical measurement instrumentation was also setup this quarter. Samples were measured and data compiled as shown in Figure 3. The most recent data points were taken at InnerSea and were consistent with data points acquired at MIT.

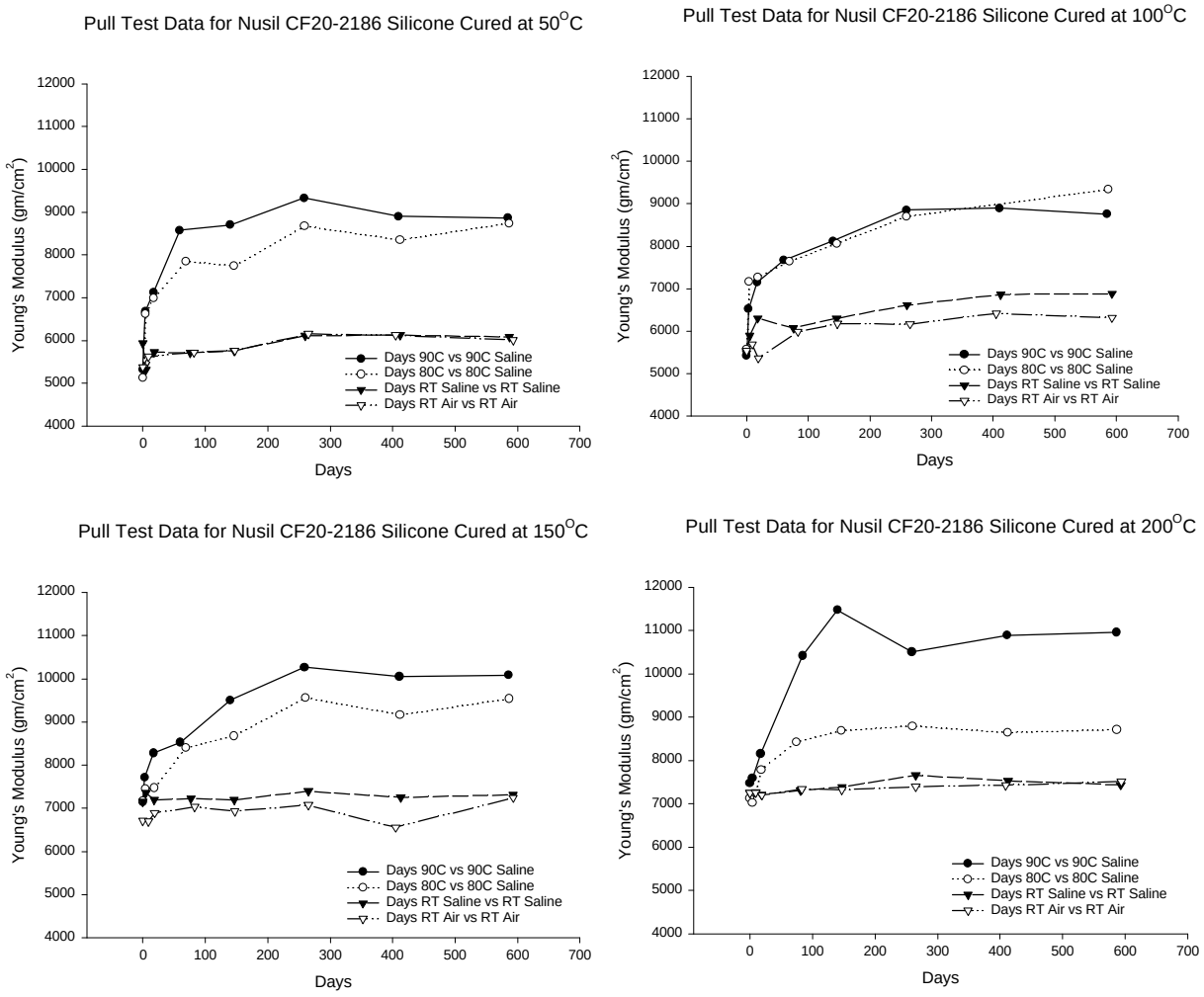

Figure 3: Young's modulus versus aging in days under various conditions for samples of Nusil CF20-2186 cured under various conditions.

Cleaning

The most important aspect of assembly of a device for implantation is cleaning of the device prior to encapsulation. Since the devices currently in use have aluminum metallization, or metallization held in place by titanium or chromium, the hot sulphuric acid/hydrogen peroxide cleans of the past cannot be used. Also, it is very difficult to avoid particulate contamination during the wire bonding operation. Thus new cleaning techniques have been continually sought. Recently, use of liquid carbon dioxide for cleaning of particulates and organic contaminants was discovered. From

demonstrations using various manufacturer's CO_2 cleaning systems, it was apparent that substantial human fingerprint contamination and particulates could be visibly removed using this technique. It was even possible to clean a bonded area of a chip without disruption of the 1 mil gold wire bonds. A system has been ordered for further investigation of this promising cleaning technique.

CMOS PassChip

The CMOS test circuit (PassChip) was further characterized and bench monitored. It was apparent from the data that significant noise existed in the system that was not attributable to the output of the PassChip. While the overall frequency content of the measurements is very low, accurate telemetry of that low frequency data requires precise control of timing. Thus the sharp edges of the data pulses must be preserved and detected at high speed. This high speed portion of the circuit was not designed to appropriately deal with the subsequent high frequency, relatively large current spikes associated with high speed switching. These large current spikes not only can couple in through the power supplies of the analog segments of the decoding circuits, but also by RF transmission and inductive coupling to sensitive nodes in the demultiplexing and filtering circuits. After substantial effort including reconfiguration of ground connections, shielding, terminating, and additional filtering, most of the switching transients were greatly reduced or eliminated. A new, multi-layer circuit board with well isolated switching and analog sections, shielded switching traces, and better decoupling of the power supplies of switching devices will be designed next quarter to eliminate altogether such noise corruption. The modified decoding system is sufficiently quiet, however, to allow continuing bench top measurements of the PassChip. An assembly substrate which will allow bonding of leadwires was fabricated by UM and will be used to assemble PassChips for saline soak testing. These substrates provide locations for attachment of leadwires and attachment of bonds from the chips to avoid direct lead wire connections to the fragile bond wires.

PPECVD Silicone

Copper wires 3 mil in diameter were coated with PPECVD silicone and placed under soak test. While the initial readings were promising, after a few days all 12 devices failed. Visual inspection revealed that failures were due to localized defects, perhaps due to particulate contamination. Additional work is needed to develop pinhole free coatings for the fine wires.

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